TITLE OF THE INVENTION

ULTRASOUND DIAGNOSTIC APPARATUS

CROSS-REFERENCE TO RELATED APPLICATIONS

This application is based upon and claims the

5   benefit of priority from the prior Japanese Patent

Application No. 11-309381, filed October 29, 1999,

the entire contents of which are incorporated herein by

reference.

BACKGROUND OF THE INVENTION

10   The present invention relates to an ultrasound

diagnostic apparatus suited for a contrast echo method

using an ultrasound contrast agent mainly composed of

microbubbles.

An ultrasound image diagnostic apparatus has many

15   advantages which other modalities, e.g., an X-ray

diagnostic apparatus, X-ray computer tomography

apparatus (CT scanner), magnetic resonance imaging

apparatus (MRI), and nuclear medicine diagnostic

apparatus (gamma camera, SPECT, and PET), do not have.

20   For example, the ultrasound image diagnostic apparatus

can generate and display images almost in real time

concurrently with scanning. This apparatus is compact

and inexpensive. In using the apparatus, there is no

chance of exposure to radiation. The apparatus can

25   easily visualize blood flow.

(Contrast agent)

Ultrasound contrast media are now being put into

practice, and an improvement in the precision of vascularity examination using such a contrast agent is expected. A contrast agent mainly consists of microbubbles of several microns. Microbubbles rapidly shrink and collapse upon reception of strong ultrasound waves. The contrast enhance effect therefore lasts only for a relatively short period of time.

In actual scan operation, since fresh contrast agent is continuously supplied to a scan slice over a blood flow, a certain degree of contrast enhance effect may be maintained. In general, however, ultrasound waves are applied several thousand times per second. In organs in which the blood flow rate is low or a relatively thin blood vessel, a contrast enhance effect can be maintained only for a moment.

(Intermittent Transmission Method)

That a contrast agent instantaneously collapses upon application of ultrasound waves is the most serious problem in the contrast echo method. As a technique of solving this problem, an intermittent transmission method is available. In this technique, scanning is intermittently repeated in synchronism with the R waves in an electrocardiogram. Fresh contrast agent continuously flows into a slice in the intervals between scans. This ensures a contrast enhance effect for every scan.

(Harmonic Imaging)

The utility of the above contrast echo method is augmented if it is used in combination with a harmonic imaging method. Microbubbles vibrate nonlinearly when ultrasound waves collide with them. This nonlinear vibrations produce harmonic components having frequencies of integer multiples of a fundamental frequency. Harmonic imaging is an imaging technique of extracting harmonic components from a fundament frequency component and visualizing them. Organs hardly vibrate nonlinearly, and hence harmonic components from the organs is pelatively small. Consequently, a region where a contrast agent exists is relatively emphasized.

As described above, if the intermittent transmission method is used, since fresh microbubbles flow into a slice during scan intervals, a contrast enhance effect can be maintained. On the other hand, in order to ensure a certain spatial resolution, about 100 to 200 ultrasound scanning lines are required per frame. That is, ultrasound pulses are applied at least a number of times equal to the number of ultrasound scanning lines in one scan.

In scanning, ultrasound pulses are transmitted while their directions are gradually changed. Therefore, microbubbles on a scanning line adjacent to a scanning line having undergone transmission/reception

may not collapse. In practice, however, since an
ultrasound beam has a certain width, most microbubbles
on the adjacent scanning line collapse in many cases.

In addition, the width of an ultrasound beam
changes in accordance with depth. The contrast enhance
effect therefore may change depending on depth.
Specific problems will be described below with
reference to FIGS. 1A and 1B.

When ultrasound pulses are generated from a
plurality of arrayed transducers 51 while the
generation timing is gradually shifted, a convergent
sound field is formed. The depth of ultrasound
focus point can be arbitrarily changed by changing
the timing. Referring to FIG. 1A, when a focus point
is formed on a scanning line with an arrow 52 at
a relatively short distance, a region exhibiting
a relatively high pressure sound, i.e., a region 53
exhibiting a relatively high degree of microbubble
collapse (microbubble collapse region), is indicated by
the hatching. The dotted lines represent adjacent
ultrasound scanning lines 54.

As shown in FIG. 1A, a short-distance focus point
is often formed by driving only transducers near
the center instead of driving all the transducers 51.
In this case, in the short-distance region, the
ultrasound beam is narrow, and hence the microbubble
collapse range 53 is also narrow. In the long-distance

region, the ultrasound beam is relatively wide,
but the ultrasound scanning line pitch is also large.
In addition, ultrasound waves greatly attenuate because
of the long propagation distance. As a result, the
5   microbubble collapse range 53 becomes relatively
narrow.

FIG. 1B shows a microbubble collapse range 55
when a focus point is formed at a relatively long
distance. As shown in FIG. 1B, to ensure a certain
10   degree of sound pressure at the focus point even at
a long distance with a long propagation distance,
many transducers must be driven at a high voltage.
In this case, in the short-distance region, the
ultrasound beam becomes wide and the microbubble
15   collapse range 55 also becomes relatively wide.
As a consequence, microbubbles on adjacent ultrasound
scanning lines collapse. When, therefore, ultrasound
transmission/reception is performed afterward in this
scanning line direction, microbubbles will have
20   collapsed and been lost.

Another problem will be described next. Consider
a case wherein a cardiac minor axis image is obtained
by the contrast echo method. As shown in FIG. 2, if
a focus point is set at a short distance in accordance
25   with a depth 61 of a cardiac muscle front wall portion
61, a rear wall portion 62 is hardly visualized due to
biological damping and a curtain effect due to

microbubbles filling a cardiac cavity.

If a focus point is set at a long distance in accordance with the rear wall portion 62, applied ultrasound pulses collapse microbubbles on adjacent

5   ultrasound scanning lines in the short-distance region. Therefore, no contrast enhance effect can be expected at the front wall portion 61.

It is known that in the sector scan method, when ultrasound waves are transmitted in a direction greatly

10   shifted from a direction perpendicular to the probe vibration surface, a decrease in sound pressure and an increase in side lobe level occur as compared with application of ultrasound wave in the direction perpendicular to the problem vibration surface.

15   That is, when ultrasound waves are applied to a cardiac muscle side wall portion 63, side lobes collapse microbubbles near the front wall portion 61 and rear wall portion 62.

It follows from the above description that it is

20   practically impossible to transmit ultrasound waves so as to form a uniform sound field distribution on an entire scan slice. The luminance reference level is therefore not uniform within the scan slice and varies. Since scattering by tissue exhibits a relatively linear

25   response, luminance can theoretically be made uniform by level correction for reception signals. However, a response of microbubbles exhibits strong nonlinearity,

and phenomena such as nonlinear vibrations, expansion, and collapse of microbubbles, vary in a complicated manner depending on the absolute level of applied sound pressure. Therefore, it is practically impossible to correct reception signals.

BRIEF SUMMARY OF THE INVENTION

It is an object of the present invention to allow an ultrasound diagnostic apparatus to exhibit a contrast enhance effect based on an ultrasound contrast agent within a slice as uniformly as possible.

An ultrasound diagnostic apparatus according to the present invention comprises an ultrasound diagnostic apparatus for obtaining an ultrasound image of a subject into which an ultrasound contrast agent mainly composed of microbubbles is injected, comprising: a probe configured to transmit/receive an ultrasound wave to/from the subject; a transmission circuit configured to drive the probe to transmit an ultrasound wave while sequentially changing a direction of an ultrasound transmission line; a reception circuit configured to generate reception line data of the number of parallel reception from ultrasound echo signals obtained by one ultrasound wave transmission; a transmission/reception control circuit configured to control the transmission and reception circuits to change the number of parallel reception during a scan sequence for generating a 1-frame ultrasound image; and

an image processing unit configured to generate an
ultrasound image on the basis of the reception line
data.

Additional objects and advantages of the invention

5   will be set forth in the description which follows, and
in part will be obvious from the description, or may
be learned by practice of the invention. The objects
and advantages of the invention may be realized and
obtained by means of the instrumentalities and combina-

10   tions particularly pointed out hereinafter.

BRIEF DESCRIPTION OF THE SEVERAL VIEWS OF THE DRAWING

The accompanying drawings, which are incorporated
in and constitute a part of the specification, illust-
rate presently preferred embodiments of the invention,

15   and together with the general description given above
and the detailed description of the preferred embodi-
ments given below, serve to explain the principles of
the invention.

FIG. 1A is a view showing a microbubble collapse

20   range when a focus point is formed at a relatively
short distance in the prior art;

FIG. 1B is a view showing a microbubble collapse
range when a focus point is formed at a relatively long
distance;

25   FIG. 2 is a view for additionally explaining
problems in the prior art;

FIG. 3 is a block diagram showing the arrangement

of an ultrasound diagnostic apparatus according to the first embodiment of the present invention;

FIG. 4 is a view for explaining parallel signal processing used in scan operation in the first embodiment;

FIG. 5A is a view for explaining the first partial scan operation in the first embodiment;

FIG. 5B is a view for explaining the second partial scan operation in the second embodiment;

FIG. 6 is a view showing the details of FIG. 5A;

FIG. 7A is a view for explaining a multifocus method used in another partial scan operation in the first embodiment;

FIG. 7B is a view for explaining the partial scan operation in the first embodiment which uses the multifocus method in FIG. 7B;

FIG. 8 is a view showing an example in which the partial scan operation in the first embodiment is applied to a linear scan;

FIG. 9 is a block diagram showing the arrangement of an ultrasound diagnostic apparatus according to the second embodiment of the present invention;

FIGS. 10A, 10B, 10C, and 10D are views showing an example of local region segmentation in the second embodiment;

FIG. 11 is a view showing another example of local region segmentation;

FIG. 12 is a view showing another example of local region segmentation;

FIG. 13 is a view showing a scan operation sequence for local regions in the second embodiment;

FIGS. 14A and 14B are views showing scan operation for local regions using a parallel signal processing method in the second embodiment;

FIG. 15 is a view showing changes in focal length with movement of a transmission beam;

FIG. 16 is a view showing an ultrasound wave application procedure using a multishot method in the second embodiment;

FIGS. 17A, 17B, and 17C are views showing a method of displaying local regions in the second embodiment;

FIG. 18 is a block diagram showing an ultrasound diagnostic apparatus according to the third embodiment of the present invention;

FIG. 19 is a view showing a scanning procedure in the third embodiment;

FIGS. 20A, 20B, 20C, 20D, and 20E are views showing an example of a combined image of luminance portions according to the third embodiment; and

FIGS. 21A and 21B are views showing display examples in the third embodiment.

DETAILED DESCRIPTION OF THE INVENTION

The preferred embodiments of the present invention will be described in detail below with reference to the

views of the accompanying drawing. Consider a clinical

case wherein an anomalous region is identified by

diagnosing the flow of blood into the liver parenchyma

or cardiac muscle by the contrast echo method.

5   (First Embodiment)

(Arrangement and Flow of Signals)

FIG. 3 shows the arrangement of an ultrasound

diagnostic apparatus according to the first embodiment.

An ultrasound probe 1 is connected to an apparatus

10   body 20. The apparatus body 20 scans the inside of

an object to be examined by using an ultrasound beam

through the ultrasound probe 1, creates tomographic

image data by processing the obtained reception signal,

and displays the image. An operating panel 9 having

15   a trackball 10A, keyboard 10B, and the like is

connected to the apparatus body 20. Various operator

instructions such as an instruction to set a region of

interest (ROI) are input to the apparatus body 20

through the operating panel 9.

20   A plurality of electroacoustic conversion elements

(transducers) are arrayed on the distal end portion

of the ultrasound probe 1. One or a few adjacent

transducers constitute one channel. RF voltage pulses

are applied from a transmitting unit 2 of the apparatus

25   body 20 to the transducers. The transducers convert

the electrical vibrations of the RF voltage pulse

into mechanical vibrations. With this operation,

the transducers generate ultrasound waves. Time
differences in the application timing of high-frequency
voltage pulses are provided between channels. These
time differences, delay times in general, are set such
5   that ultrasound waves generated by the transducers are
combined into one narrow beam, and the beam is
deflected as needed. By changing the delay times, the
focal length and deflection angle can be arbitrarily
changed. A transmission/reception control circuit
10   (T/R) 13 controls these delay times.

Ultrasound waves propagate through the object and
are reflected by an acoustic-impedance discontinuous
surface located at some point in the object. The
reflected waves return as echoes to the probe 1.
15   The echoes mechanically vibrate the transducers of the
probe 1. As a consequence, weak current signals are
generated. A receiving unit 3 amplifies the current
signals in units of channels, converts them into
voltage signals, and converts them into digital
20   signals. In addition, the receiving unit 3 adds the
signals while giving them delay times that differ
between the channels. This addition is processing
called digital beam forming, by which a reception
signal is given a directivity. The receiving unit 3
25   has a plurality of digital beam forming systems.
The transmission/reception control circuit (T/R) 13
simultaneously generates a plurality of reception

signals having different directivities by parallel processing and differently controlling delay times between the digital beam forming systems.

The transmission/reception control circuit 13 implements scan operation (to be described later) according to the present invention by controlling delay times in transmission and reception. Note that scan operation is defined as operation to acquire a plurality of reception signals required for the creation of a 1-frame image. More specifically, scan operation is operation of repeating a series of ultrasound beam transmitting/receiving operations with respect to a plurality of ultrasound scanning lines constituting a scan plane. This new scan sequence makes it possible to obtain both the effect of improving the time resolution and the effect of making the contrast enhance effect of microbubbles relatively uniform within a scan plane.

In addition to the transmission/reception delay control function, the transmission/reception control circuit 13 has basic control functions such as the function of setting a transmission frequency and the function of shaping the waveform of a transmission pulse. As described above, by changing the delay times in transmission/reception, the transmission direction, reception direction, focal length, and ultrasound scanning line density of ultrasound beams can be

arbitrarily changed. In general, parameters such as delay time and transmission frequency differ between modes such as the B-mode and color Doppler mode. B-mode and color mode data can be simultaneously obtained by alternately transmitting these waves.

A receiver 4 and subsequent components will be described next. The receiver 4 is comprised of a logarithmic amplifier, an envelope detection circuit, a band-pass filter for extracting harmonic components from a reception signal, and the like.

An output from the receiver 4 is converted by a B-mode DSC 5 from a fan array of ultrasound scanning lines into an orthogonal array of scanning lines corresponding to a standard video format. The resultant data is sent as a bit stream to a combination circuit 6. The combination circuit 6 combines image data and additional information such as an electrocardiographic waveform and various set values into one frame, thereby forming a frame to be finally displayed on a display 7.

A memory control circuit 14 sends array conversion information to the B-mode DSC 5 and combination circuit 6.

An image memory 8 temporarily stores the signal train after the array conversion by the B-mode DSC 5 (or the signal train before the array conversion). This information is read out by an operator after

a diagnosis or the like. In this case, the information
is output to the display 7 through the B-mode DSC 5 and
combination circuit 6.

An ECG analyzer 12 analyzes the electrocardiograph
5 (ECG) data of the object measured by an ECG 11,
extracts, for example, R waves, and generates a trigger
signal to the transmission/reception control
circuit 13. The ECG analyzer 12 converts the
electrocardiograph data into display data and sends it
10 to the combination circuit 6. This electrocardiograph
data and tomographic image data are combined into
a single frame to be displayed together on the
display 7. A clock 15 is used to control intermittent
transmission intervals in a diagnosis using on ECG
15 signal, e.g., a diagnosis of an abdominal organ.
Note that the operator can control the intermittent
transmission intervals and timing on the operating
panel, and the control operation is reflected in the
transmission/reception control circuit 13.
20 (Scan Operation)

The scan operation in this embodiment will be
briefly described by exemplifying the parallel signal
processing of simultaneously generating a plurality of
reception signals having different directivities for
25 one transmitting operation. Consider a case wherein
two reception signals having different directivities
are obtained for one scanning operation. FIG. 4 is

a view for explaining the principle of parallel signal

processing. Referring to FIG. 4, "r" represents

an ultrasound scanning line; and "t", the transmission

direction. Ultrasound pulses are transmitted under

5   delay control corresponding to the direction of

an ultrasound scanning line t1. From the resultant

echo signals, digital beam formers of two systems

generate two types of reception signals that are

given directivities in the directions of ultrasound

10   scanning lines r1 and r2 under two types of delay

control. A reception scanning line density twice

a transmission scanning line density is realized

by this parallel signal processing. Obviously,

four or more reception signals with different

15   directivities can be theoretically generated for

one transmitting operation.

The scan operation in this embodiment will be

described in detail next with reference to FIGS. 5A

and 5B. Assume that one scan plane consists of 160

20   ultrasound scanning lines. The angle difference

between adjacent ultrasound scanning lines is

represented by θ.

Scan operation for acquiring reception signals

required to generate a 1-frame image is constituted by

25   a plurality of, two in this case, partial scan

operations. More specifically, in the first partial

scan operation, the first portion in the scan region is

scanned, and in the second partial scan operation,
the remaining, second portion in the scan region is
scanned. A partial image of the first portion obtained
by the first partial scan operation and a partial image
5   of the second portion obtained by the second partial
scan operation are combined into one frame, thereby
completing a 1-frame image of the entire scan plane.

In the first and second partial scan operations,
the focus point is fixed to a long distance. FIG. 5A
10 shows the first partial scan operation. FIG. 5B shows
the second partial operation. As shown in FIG. 5A,
ultrasound transmission/reception is repeated in
predetermined cycles. A transmission beam is
sequentially moved from the right end to the left
15 end of a scan plane in each transmission/reception.
The intervals at which a transmission beam is moved are
set to an integer multiple equal to or more than two of
the angle difference θ between ultrasound scanning
lines, $4 \cdot \theta$ in this case.

20 In digital beam forming, a plurality of, four
in this case, reception signals having different
directivities are generated in every transmission by
parallel signal processing. More specifically, four
reception signals having directivities corresponding to
25 fourth ultrasound scanning lines which are symmetrical
about a transmission beam are generated.

These four reception signals are subjected

to detection and luminance conversion in the receiver 4

and written in the B-mode DSC 5. Different

write sequences for these signals are used in

a short-distance region A and long-distance region B.

5   In the short-distance region A, all the four reception

signals are written. In the long-distance region B,

only the two reception signal corresponding to the two

ultrasound scanning line located in the center are

written.

10   This first partial scan operation will be

described in detail below with reference to FIG. 6.

FIG. 6 shows a microbubble collapse range 100 in which

a high contrast enhance effect is obtained when the

focus point is formed at a relatively long distance.

15   The long-distance focal point is formed under a typical

transmission condition, i.e., a wide aperture and high

driving voltage. In this condition, the width of the

microbubble collapse range 100 is wide at the

short-distance region A and narrow at the long-distance

20   region B, as described with reference to FIG. 1B. The

contrast enhance effect is high within the microbubble

collapse range 100.

In accordance with this region exhibiting a high

contrast enhance effect, in the short-distance region

25   A, all four reception signals (luminance signals)

corresponding to four ultrasound scanning lines r1, r2,

r3, and r4 which are symmetrical about a transmission

beam t11 are written in the B-mode DSC 5. In the long-distance region B, only the two reception signal corresponding to the two ultrasound scanning lines located in the center are written in the B-mode DSC 5.

The transmission beam is them moved to t12, and all four reception signals (luminance signals) corresponding to four ultrasound scanning lines r5, r6, r7, and r8 which are symmetrical about the transmission beam t12 are written in the B-mode DSC 5. In the long-distance region B, only the two reception signals corresponding to the two ultrasound scanning lines located in the center are written in the B-mode DSC 5.

When the first partial scan operation is completed upon repeating this sequence, on the memory of the B-mode DSC 5, the entire short-distance region A is filled with the luminance data, and the long-distance region B partially has luminance data blank portions. These blank portions are filled with the luminance data obtained by the second partial scan operation.

Disregarding averaging, the number of times of transmission/reception required for the first partial scan operation is 160/4 = 40.

As is obvious, since only the reception signals within the microbubble collapse range 100 need be finally used for an image, unnecessary reception signals need not be generated in the first partial scan

operation, i.e., the reception signals corresponding to the two outside ultrasound scanning lines in the long-distance region B need not be generated by digital beam forming.

The second partial scan operation will be described next. In the first partial scan operation, data blank portions are present at two adjacent ultrasound scanning lines in the long-distance region B. Referring to FIG. 5A, for example, ultrasound scanning lines r4 and r5 correspond to blank portions. In the second partial scan operation, transmission, digital beam forming, and DSC write operation are performed to fill these blanks with data, as shown in FIG. 5B.

Ultrasound transmission/reception is repeated in predetermined cycles like the first partial scan operation. A transmission beam is sequentially moved from the right end to the left end of a scan plane in every transmitting/receiving operation. The intervals at which the transmission beam is moved are set to $4 \cdot \theta$ as in the first partial scan operation. However, the transmission beam in the second partial scan operation is shifted from the transmission beam in the first partial scan operation by half the moving intervals of the transmission beam, i.e., $2 \cdot \theta$. With this operation, an ultrasound beam is transmitted in the direction of the center axis of each ultrasound

scanning line corresponding to a blank portion formed
in the first partial scan operation. The first
transmission beam in the second partial scan operation
is transmitted to a direction t21 between blank

5   scanning lines r4 and r5. The next transmission beam
is transmitted in a direction t22 between blank
scanning lines r8 and r9. In this manner, transmission
is repeated at intervals of $4 \cdot \theta$.

     In digital beam forming, a plurality of, two in

10   this case, reception signals having different
directivities are generated by parallel signal
processing for each transmitting operation.
More specifically, two reception signals having
directivities corresponding to left and right

15   ultrasound scanning lines on the two sides of
a transmission beam are generated.

     These two reception signals are subjected to
detection and luminance conversion in the receiver 4,
and only the data corresponding to the data blank

20   portions in the long-distance region B formed in the
first partial scan operation are written in the B-mode
DSC 5.

     As described above, in the second partial scan
operation, image portions are assembled such that they

25   do not overlap the image portions generated in the
first partial scan operation in FIG. 5A, and no blank
portions are formed. The number of times of

transmission/reception in FIG. 5B is set to 40 (39, to

be exact) as in the case shown in FIG. 5A. That is,

transmission/reception is performed a total of 80

times, which is equal to the number of times of

5 transmission/reception in 2-direction parallel signal

processing. That is, the frame rate does not decrease.

The merits of this scan operation will be

described below. At a long-distance focus point, the

beam width increases in the short-distance region A.

10 In this short-distance region A, microbubbles collapse

in a wide range. In other words, in the short-distance

region A, a contrast enhance effect owing to

microbubbles can be obtained in a wide range. This

makes it possible to effectively generate reception

15 signals at once from a wide range corresponding to as

many as four scanning lines for one transmitting

operation by 4-direction parallel signal processing.

On the other hand, in the long-distance region B, the

intervals between ultrasound scanning lines become

20 larger than those between ultrasound scanning lines in

the short-distance region A, and the distance from the

center of a transmission beam increases, resulting in

a decrease in the intensity of ultrasound waves.

If 4-direction parallel signal processing is performed

25 in this state, sensitivity deteriorates. For this

reason, in the first partial scan operation,

2-direction parallel signal processing is performed.

In the second partial scan operation, then, only
the blank regions in the first partial scan operation
are filled with data. That is, the blank regions in
the first partial scan operation correspond to data

5   on scanning line pairs in the long-distance region.
At a long-distance focus point, the scanning lines in
these blank regions are included in the microbubble
collapse range exhibiting a high contrast enhance
effect. In this manner, high contrast enhance effects

10   can be ensured in both the long- and short-distance
regions.

In this case, for the sake of simplicity, the
short-distance region A and long-distance region B
are separated from each other by a clear boundary.

15   To obtain a smoother image, these regions may overlap.
In this case, for example, the luminance image on the
overlapping portion is averaged to make the boundary
less noticeable.

This scan operation is especially effective

20   for sector scan operation in which ultrasound scanning
line intervals differ in a short-distance region and
long-distance region. However, effects similar to
those described above can also be obtained when this
operation is applied to linear scan operation in which

25   ultrasound scanning line intervals remain unchanged in
a short-distance region and long-distance region.

In addition, this technique may be used in

combination with the intermittent transmission method
of performing transmission in synchronism with an ECG
signal. In this case, since the application of
ultrasound waves is stopped during periods other than

5   synchronous periods, more microbubbles flow into
a slice of a region of interest and are stored without
collapse. Obviously, if ultrasound waves are applied
in this state, more microbubbles can be detected.
In addition, if the scanning method of this embodiment

10   is used, the contrast enhance effect can be relatively
improved.
(Other Examples of Scanning Method)
       A multifocus method is available as a conventional
method similar to the scanning method to be described

15   below. The multifocus method is currently implemented
in many apparatuses. This method will be described
first with reference to FIGS. 7A and 7B. According to
the multifocus method, as shown in FIG. 7A, (a) the
respective ultrasound scanning lines are scanned to

20   perform transmission/reception at a short-distance
focus point so as to generate a signal component
corresponding to a short-distance region A, and (b)
a signal component corresponding to a long-distance
region B is generated at a long-distance focus point.

25   These two images are then combined into a 1-frame
tomographic image. Each ultrasound scanning line
has two focus points (represented by the heads of

the arrows in FIG. 7A), and hence the resolution

increases. However, since the number of times of

transmission/reception increases twice, the frame rate

decreases to half. In addition, a multifocus method

5   using three or more focus points is also available.

In the above method, if the position of the focus

point is sequentially changed from the short-distance

region, microbubble echoes may be detected up to

a deep portion while microbubbles collapse from the

10   short-distance region. However, microbubbles on

adjacent ultrasounds collapse.

The second example of the scanning method of this

embodiment, which aims at solving the above problem,

will be described below with reference to FIG. 7B.

15   Note that the bullets on the ultrasound scanning lines

represent focus points, and the numbers on the lower

portion of the drawing represent a scan sequence.

As is obvious from FIG. 7B, a characteristic feature of

this scanning method is that an ultrasound scanning

20   line r2 is scanned first at a short-distance focus

point, an ultrasound scanning line r1 immediately

preceding the ultrasound scanning line r2 is then

scanned at a long-distance focus point, and scanning is

performed in the order of r4, r3, r6, r5, ... In this

25   manner, while the scanning beam moves forward from one

set of a plurality of, two in this case, adjacent

ultrasound scanning lines to another set, the beam

moves backward in the transmission direction at each

set. In addition, focus points are alternately

switched in the short-distance region and long-distance

region in each transmission, thereby obtaining the

5 following effects.

Assume that the ultrasound scanning lines are

sequentially scanned from the ultrasound scanning line

r1 as in a conventional method. In this case, when an

ultrasound pulse is transmitted to the ultrasound

10 scanning line r1, microbubbles on the ultrasound

scanning line r2 are affected, e.g., collapse.

However, since the ultrasound scanning line r2 is

scanned first, this adverse effect can be avoided.

In addition, since the ultrasound scanning line r2 is

15 scanned at a short-distance focus point, the influence

of this scanning on the next ultrasound scanning line

r1 is small. That is, microbubbles on the ultrasound

scanning line r1 do not collapse much. As described

above, this method can minimize the collapse of

20 microbubbles on adjacent ultrasound scanning lines due

to the application of ultrasound waves.

Note that if different focus points are set on the

respective ultrasound scanning lines as in this case,

since different sound fields are formed, echo signals

25 on adjacent ultrasound scanning lines may be made

uneven. Smoothing by, for example, averaging on

adjacent ultrasound scanning line to reduce such

unevenness is effective for an improvement in image
quality.

Although the scanning method of the above
embodiment is a scanning method represented by a sector
probe, the present invention can also be applied to
a linear type scanning method. As shown in FIG. 8, in
the linear type scanning method, the intervals between
ultrasound scanning lines do not depend on depth.
However, the profile of the sound field formed by one
beam is the same as that in the above method, and the
influences on microbubbles on adjacent ultrasound
scanning lines in a short-distance region still remain.
When the present invention is to be applied to the
linear type scanning method, ultrasound scanning lines
are formed to be spaced from each other by a distance
4d, and the ultrasound scanning lines are shifted by
a distance 2d to form ultrasound scanning lines spaced
part from each other by the distance 4d. In this case,
the time required to generate one frame becomes
equal to that in the scanning method in which the
transmission ultrasound scanning line density is 2d.

This embodiment presents a scanning procedure
by which signals derived from microbubbles can be
efficiently received, and the unevenness of contrast
enhance effect within a slice can be corrected when one
tomographic image is to be generated by the contrast
echo method performed by administering a contrast

agent. With this procedure, even if an ultrasound
contrast agent exhibiting the same performance as that
of a conventional contrast agent is administered,
the contrast enhance effect can be relatively improved.

5   Therefore, an improvement in blood flow diagnosis
ability, especially an improvement in fine blood flow
diagnosis ability, is expected.

(Second Embodiment)

The second embodiment provides a partial imaging

10   method. The partial imaging method is a method of
segmenting a scan plane into a plurality of local
portions and scanning each local portion in an optimal
scan operation sequence, instead of sequentially moving
over scanning lines within the scan plane, thereby

15   obtaining optimal (maximum) contrast on the entire scan
plane and combining the resultant data into a one
frame.

(Arrangement and Flow of Signals)

FIG. 9 is a block diagram showing an ultrasound

20   diagnosis apparatus of this embodiment. The same
reference numerals as in the first embodiment denote
the same parts in the first embodiment, and a detailed
description thereof will be omitted. A template memory
21 stores pieces of information about a plurality of

25   models for segmenting a scan slice into a plurality of
local regions (partial regions). One optimal pattern
for a sliceal shape of a portion to be diagnosed is

read out from the template memory 21 in accordance with an instruction input by the operator on an operating panel 9. This template is sent to a memory control circuit 14 first, and then displayed on a display 7 in a form in which the template is superimposed on an ultrasound diagnosis image. A signal processing unit 22 performs numeric operation such as averaging echo signals from a local region or obtaining a representative value.

A graphic unit 23 performs graphic signal processing, e.g., synthesizing an image on the basis of data sent from the signal processing unit 22 and coloring a simplified graphic pattern. The image data created by this processing is output to the display 7 through a combination circuit 6. This data is also transferred to an external computer, printer, or the like through a network board 24.

(Region Segmentation in Partial Imaging Method)

This partial imaging method is roughly constituted by three steps, i.e., (a) the step of setting a diagnosis region of interest and a local region of interest, (b) the scanning step, and (c) the display step.

(a) Setting Diagnosis Region of Interest and Local Region of Interest

First of all, it is important to set a specific local portion toward which the diagnostic apparatus is

to perform optimal scanning. The settings vary
depending on an organ to be diagnosed and its slice.
For example, the cardiac muscle in a cardiac minor axis
image has an almost circular shape. A circular
template 50 representing a local region like the one
shown in FIG. 10A is selected in advance. The size of
the template 50 is then adjusted by using the zoom
function of the operating panel 9 or the like to almost
overlap the outer ring of the template 50 on the minor
axis image, as shown in FIG. 10B. If the size of
a template is fixed in accordance with a routine,
the template need not be displayed in some case.

As shown in FIG. 10C, the operator then designates
desired local regions with points or regions.
Obviously, a plurality of regions can be designated.
In this case, regions can be set at uneven intervals
along the cardiac muscle. As shown in FIG. 10D,
the operator designates representative points of
those set above. With this operation, the apparatus
performs automatic segmentation. In this example,
diametrically located four points are designated to
obtain a segmented region like the one shown in
FIG. 10B. Obviously, the size of a template,
the number of local regions segmented, and the like
can be changed.

In the above case, the cardiac minor axis image is
exemplified. However, a template suited for the shape

or the like of another slice, such as a 2-cavity cross

section or major axis image, may be selectively used.

In addition, the setting method shown in FIGS. 10C

and 10D can also be used. In diagnosing the liver,

5   since the liver is included in an overall slice,

relatively simple local region segmentation like

the one shown in FIG. 11 can be performed.

(b)   Scan Operation

The set local regions are sequentially scanned.

10   In general, this scanning is started when the operator

presses a start button on the operating panel 9.

In the case shown in FIG. 12, for example, when

a local region A is to be irradiated, a focus point

is set at the central portion of the region A.

15   A transmission/reception control circuit 13 changes the

ultrasound transmission conditions in accordance with

the position of each local position (transmission focus

point) so as to almost equalize the degrees of dynamic

influences on the respective local regions, i.e., sound

20   pressures on the respective local regions and the

degrees of collapse of microbubbles in the respective

local regions. Typical transmission conditions that

can be adjusted include the driving voltage for each

transducer, the aperture width (the number of

25   transducers to be simultaneously driven), the driving

frequency, and the scanning line density.

In this case, the respective parameters are

changed to almost equalize the degrees of collapse of
microbubbles in consideration of biological damping
(mainly determined by the transmission distance of
ultrasound waves) and the irradiation angle with
5   respect to each transducer. When the parameters
associated with the driving voltage for each transducer
and the number of transducers to be simultaneously
driven are changed, the degrees of collapse of
microbubbles can be made almost uniform by almost
10   equalizing the sound pressures at the positions of the
respective focus points. More specifically, when the
driving voltage for each transducer is to be changed,
the driving voltage is lowered if the focus point is
located near, and vice versa. When the number of
15   transducers to be driven is to be changed, the number
of transducers to be driven is decreased if the focus
point is located near, and vice versa. When the
frequency is to be changed, the frequency is increased
if the focus point is located near, and vice versa.
20   Note that only one of the above parameters may be
changed or a plurality of parameters may be
simultaneously changed.

Strictly speaking, the value of damping varies
among objects to be examined. However, a rough value
25   can be presented to the operator on the basis of data
obtained in advance by measurement. With reference to
a region E exhibiting the maximum damping, relative

transmission sound pressures for the respective regions
are set such that -1.5 dB is set for regions D and
F; -2 dB, for regions C and G, -4.5 dB, for shallow
regions B and H, and -6 dB, for a shallowest region A.

5   If an effective echo signal is to be obtained by
irradiating the local region E with ultrasound waves,
the region A on the same ultrasound scanning lines is
affected by the collapse of microbubbles. It is
therefore useless to scan the region A immediately
10   after the region E. This problem can be effectively
solved by using an intermittent transmission method.

As shown in FIG. 13, for a circulatory organ
region, transmission is performed once for every
heartbeat or for a few heartbeats by using triggers
15   synchronous with an electrocardiogram. In the interval
between triggers, no transmission is performed, and
hence new microbubbles flow into a region of interest
upon collapse of microbubbles. This makes it possible
to acquire a sufficient contrast enhance effect again.

20   In the case shown in FIG. 13, at the same trigger
timing, echo signals are generated from the pairs
of local regions B and H, C and G, and D and F at
the same heartbeat timing by using the parallel signal
processing method. This is because each pair of
25   regions are located at the same depth and same focal
length, and do not affect each other, i.e., do not
collapse microbubbles each other, since they are spaced

apart from each other. In addition, in the case shown
in FIG. 13, scanning from the region A to the region H
is repeated a plurality of number of times.

FIGS. 14A and 14B show another operation

5   procedure. Assume that scanning is started from the
right end of the screen. In this case, at the first
trigger, as shown in FIG. 14A, the local regions A, B
and C are simultaneously scanned by using 3-direction
parallel signal processing. In each region, however,

10   the focal length and output sound pressure can be
changed, and the changed values are set as optimal
values in each region. At the second trigger, as shown
in FIG. 14B, similar scanning is performed for regions
on which the ultrasound scanning lines at the first

15   trigger overlap. In this method, the segmentation
forms of local regions are not uniform unlike those in
FIG. 11. A merit of the method is that microbubble
echoes can be obtained relatively effectively with
a smaller number of times of transmission/reception.

20   FIG. 15 shows an advanced method. At the first
trigger, the focal length changes for each ultrasound
scanning line (an illustration of ultrasound scanning
lines is omitted), as indicated by "A" in FIG. 15.
The output sound pressure is also controlled in

25   accordance with a change in focal length such that
output pressures on other ultrasound scanning line in
this focus point portion are made uniform. At the

second trigger, the focal length changes for each

ultrasound scanning line as indicated by "B" in

FIG. 15. As a result, at the second trigger, good

echoes can be received from the entire circumferential

5   portion of the cardiac muscle, and echoes can be

acquired under a uniform sound pressure intensity along

the circumferential portion of the cardiac muscle.

The following technique can also be used to

extract clearer microbubble echoes. According to this

10   technique, as shown in FIG. 16, immediately after

an echo signal is obtained by performing a scan T1 for

a given region, a similar scan T2 is performed for the

same region at the above trigger timing. If the scan

T2 is performed at this timing, microbubbles collapse

15   by the immediately preceding scan T1, and only an echo

signal representing a tissue remains in the echoes

obtained by the immediately succeeding san T2. If the

receiver calculates the difference between the echo

signals obtained by these two scans, only the signal

20   derived from the collapsed microbubbles is extracted as

a difference signal. As a consequence, the echoes

derived from the contrast agent which are free from

the influences of luminance of the living tissue can

be visualized (this method will be referred to as

25   a subtraction method hereinafter).

(c) Display

According to the data acquisition method of the

present invention described above, since data (signal
intensity) presenting a contrast enhance effect on the
entire region of interest can be obtained by executing
at least two scan procedures, the technique of

5   combining the data and displaying the resultant image
is used. In the case shown in FIG. 12 or 14, since the
boundaries between the respective local regions are
clear, a simple image (simple signal intensity
distribution) is reconstructed by the combination

10   circuit 6 using only data from a corresponding region,
and the image is displayed on the display 7.

This image combining and displaying method can be
applied as follows. In diagnosing a myocardial blood
flow, attention is often paid to find an ischemic

15   region of the cardiac muscle. In this case, the
operator need only know fine blood flow perfusion due
to a contrast agent without paying any attention to
the fine form of speckle pattern of the cardiac muscle
(as in the case of a scintigram in nuclear medicine).

20   To meet such needs in diagnosis, for example, a display
method of calculating the average contrast degree
of each local region and displaying the calculated
value as the representative value of each local
region is used. Painting each region by using color

25   information based on a color bar or the like makes
the displayed information easier to discern (FIG. 17A).
Alternatively, simplified display may be performed as

shown in FIG. 17B. Furthermore, as shown in FIG. 17C, blood flow rates may be numerically expressed and displayed. Although the luminances and numerical values in display represent relative information, such

5   a display method allows the operator to quickly detect an ischemic region of the cardiac muscle if it exists. These images may be displayed side by side as well as being overlaid on an original diagnostic image. More simplified display images can be added to

10   a patient's chart or electronic patient's chart through a means such as a network as well as being output from a printer.

(Third Embodiment)

FIG. 18 shows the arrangement of the third

15   embodiment. A transmission/reception control circuit 13 controls the timing of pulses from a transmitting unit. This embodiment performs intermittent transmission using a timing signal from an ECG analyzer 12 or clock 15 in accordance with a mode

20   switching instruction sent from an operating panel 9. Intermittent transmission is transmission in which frame generation intervals are sufficiently larger than those in normal continuous transmission (20 to 100 frames/sec). For example, time intervals

25   corresponding to four or five heartbeats are input.

The transmission/reception control circuit 13 instructs a transmitting unit 2 to perform transmission

of a plurality of frames per trigger (in other words,
continuous scanning in a short period of time).
FIG. 19 shows a conceptual rendering of this
transmission. Referring to FIG. 19, five frames are

5   continuously transmitted/received at predetermined time
intervals (sufficiently longer than the frame rate).
An image processing unit 31 combines a plurality of
frames obtained per trigger by the above method and
sends the resultant images to the combination circuit,

10   and the images are displayed on the display. Prior to
a description of an image processing method, diagnostic
images expected from the transmission method in FIG. 19
will be described.

<Problems to Be Solved>

15   If the contrast agent concentration is relatively
low or a contrast agent is made of microbubbles that
easily collapse, most microbubbles on a slice collapse
by a transmission pulse in the first frame. In the
second and subsequent frames, therefore, diagnostic

20   images are formed without any microbubbles, i.e., made
of only tissue echoes.

It is, however, empirically known that there are
cases other the above case. If, for example, the
contrast agent concentration is high or a recently

25   developed contrast agent containing microbubbles that
are relatively resistant to ultrasounds is used, the
following phenomenon occurs.

As in FIG. 20A, in transmission of the first
frame (A), a contrast enhance effect is seen at
a relatively short-distant portion. However,
the collapsing effect of high-concentration
5   microbubbles at the short-distance portion increases,
and hence ultrasound pulses can hardly propagate to
a deeper portion. As a consequence, no image is seen
at a portion deeper than the short-distance portion,
and the portion becomes dark. In some case, such
10   a portion becomes darker after administration of
the contrast agent than before (called shadowing).
As shown in FIG. 20B, in the second frame (B), since
the microbubbles at the short-distance portion have
collapsed, the contrast enhance effect decreases.
15   However, since damping of sound waves due to
microbubbles is reduced, a relatively deep portion
is irradiated with a relatively high sound pressure.
The contrast enhance effect at this portion increases.
Subsequently, a similar phenomenon is transferred to
20   deeper portions. When microbubbles collapse in all
regions as shown in FIG. 20C, the displayed image is
formed by tissue echoes (FIGS. 20D and 20E).

If this phenomenon is seen with a moving picture,
the movement of the luminance based on contrast from a
25   shallow portion to a deep portion like a curtain that
drops is recognized. This phenomenon will be referred
to as a "curtain phenomenon" hereinafter.

Obviously, when the above phenomenon occurs, the operator cannot examine a contrast enhance effect on an entire slice by only seeing one of a plurality of images. All the images must be joined to each other.

5 <Arithmetic Image Processing>

The image processing unit 31 described above performs optimal image combining processing when the above curtain phenomenon occurs. More specifically, images obtained at one trigger timing are stored in

10 the image memory 8. The luminance signals of these images at the same coordinates in the respective frames are compared with each other to detect maximum values. Arithmetic processing for determining a luminance $I(x, y)$ at coordinates (x, y) is given by

15 $$I(x, y) = MAX(Ii(x, y)), \qquad I = 1 \cdots N$$
where $I_i(x, y)$ is the luminance at the coordinates (x, y) in the ith frame, and N is the number of images to be compared with each other.

As a result of this processing, images like those

20 shown in FIGS. 20A, 20B, 20C, 20D, and 20E having high-luminance contrast portions joined together are combined, and the combined image is displayed on the display. As is obvious from the result, since this image displays the luminance corresponding to the

25 highest contrast among all the regions, the operator can examine the overall contrast degree with this one diagnostic image.

Note that this technique is similar to an MIP (Maximum Intensity Projection) method used to project three-dimensional space information on a two-dimensional plane. However, the general MIP method is used for spatial points, whereas the technique of the present invention is used for temporal points.

Note that the above arithmetic processing is relatively simple, and a combined image is preferably displayed almost in real time immediately after transmission at the trigger timing.

As described above, this embodiment cannot exhibit a sufficient effect when no curtain phenomenon occurs, but has no adverse effect. Therefore, this technique is not used in any specified condition.

<Examples of Display>

FIGS. 21A and 21B show examples of a display form. FIG. 21A shows a method based on two-window display. While intermittent transmission is observed in real time on one window, the above combined image is sequentially displayed on the other window. Referring to FIG. 21B, images obtained at one trigger timing are displayed side by side, and a combined image is simultaneously displayed. Note that all image need not have the same size. In general, since a combined image is most important for diagnosis, the image is preferably displayed in a relatively large size, as shown in FIG. 20B.

Additional advantages and modifications will readily occur to those skilled in the art. Therefore, the invention in its broader aspects is not limited to the specific details and representative embodiments

5   shown and described herein. Accordingly, various modifications may be made without departing from the spirit or scope of the general inventive concept as defined by the appended claims and their equivalents.